A new multi-directional fiber model for low angular resolution diffusion imaging

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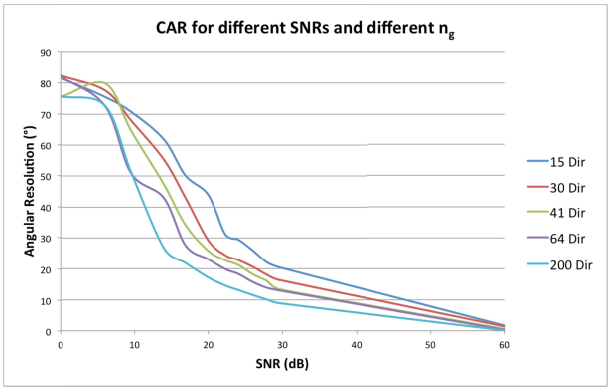
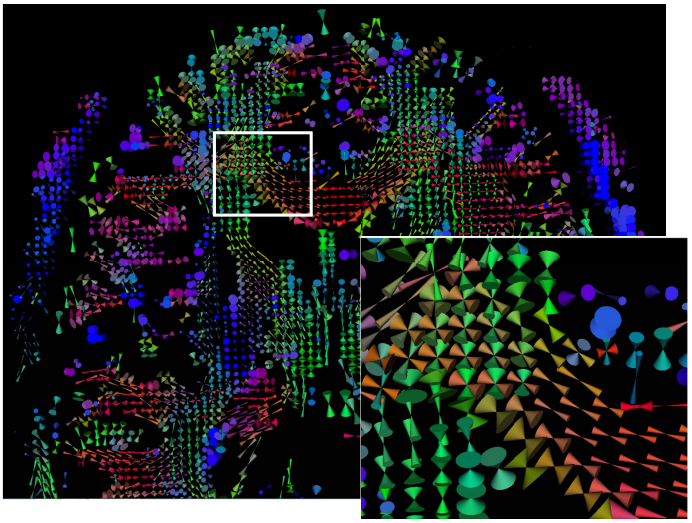
Introduction: Diffusion MRI (dMRI) permits to infer the ensemble average propagator (EAP) from a set of diffusion-weighted (DW) images acquired from n_g gradient directions and n_b b-values. In the context of clinical brain imaging, dMRI sequences seldom exceed 10 minutes acquisition, with $n_g \leq 30$ and only one b-value. The EAP is then inferred from the resulting low angular resolution diffusion (LARD) images by assuming a Gaussian diffusion profile [1]. In research context, higher angular resolution samplings ($n_g \geq 60$ and $n_b \geq 1$) [2,3,4] have revealed a non-Gaussian diffusion profile in the white matter when fibers cross. To account for that effect, we propose a non-Gaussian parametric modeling of the EAP, the estimation of which can be accurately performed from LARD images obtained in clinical context.

Theory: In each voxel, the EAP is modeled as a mixture. Each probability density function (pdf) in the mixture characterizes the diffusion along some fiber orientation (FO) $\pm\boldsymbol{\mu}$, $||\boldsymbol{\mu}||=1$, and is in turn modeled as a mixture of two equally weighted pdfs that account for the diffusion along directions $\boldsymbol{\mu}$ and $-\boldsymbol{\mu}$ respectively. The diffusion pdf along direction $\boldsymbol{\mu}$ is given by the convolution of a von Mises & Fisher pdf on the sphere of radius R>0 (mean covered distance), with mean direction $\boldsymbol{\mu}$ and concentration (around $\boldsymbol{\mu}$) parameter $\kappa \geq 0$, and a centered 3D Gaussian pdf with covariance matrix $D=R^2 (I_3+\kappa\boldsymbol{\mu}\boldsymbol{\mu}')/(\kappa+1)$ (cylindrical shape). Crossing fibers are consequently characterized by 8 parameters. The Fourier transform of the EAP is analytically derived as a function of the parameters of the model and yields the theoretical DW intensity [5]. The estimation of these parameters is then performed by a least squares fitting of the observed DW intensities to the theoretical ones.

Methods: An evaluation of the crossing angle resolution (CAR) of the model was first performed using synthetic data on a single voxel. These data were generated as in [6] for different configurations of the two FOs with b = 1500s/mm^2 and n_g=15, 30, 41, 64 and 200. The resulting data sets were then corrupted with increasing Rician noise and, for each noise level σ, 100 samples were synthetized. For a given n_g and σ, the CAR was computed as the 95% confidence angle between the two estimated FOs in situations where the real FOs are collinear. A healthy adult male was scanned on a 3T Achieva Philips MRI Scanner with a 8-ch head coil, TR/TE/τ = 10000/64/22.1ms, b=800s/mm^2, n_g=15 and 2x2x2mm^3 voxels. This set of DW images represents a typical case of LARD images with low spatial resolution from which our model of the EAP was estimated.



Results: Figure on the left shows the CAR of the model for increasing signal-to-noise ratios (SNR). Each curve corresponds to a given n_g. For low SNRs, increasing n_g does not significantly improve the CAR. For typical clinical values of SNR = 20dB and n_g=30, the corresponding CAR of 30° outperforms the CAR obtained in Q-Ball Imaging [7], i.e. around 60° for higher angular resolution (n_g=81) [8]. Figure on the bottom shows an extremity of the corpus callosum known to contain crossing fibers (the height of the cones is proportional to R^2 while the radius is proportional to $1/(\kappa+1)$). Fiber crossings seem to be accurately estimated despite the low angular and spatial resolutions.



Discussion: This model enables crossing fibers to be theoretically estimated from only 8 DW images. In particular, this model allows for the retrospective study of DW data sets acquired over the past years. For a complete applicability in clinics, one could wonder whether maps akin to the fractional anisotropy (FA) and mean diffusivity (MD) maps [9] can be provided with this model. For a given FO, based on the Gaussian part of our model and by analogy with DTI, we propose $FA = \kappa((\kappa+1)^2+2)^{-1/2}$ and $MD = (1+\kappa/3)\, R^2/(1+\kappa)$.

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